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TITLE: The influence of single-leg landing direction on lower limbs biomechanics

RUNNING HEAD: Single-leg landing tasks profile

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Abstract

Background: The purpose of this study was to investigate whether different directions of single-leg landing tasks would influence lower limbs joints biomechanics.

Methods: Using a controlled and experimental design, thirty physically active male subjects (age: 26.4 ± 4.4 years; height: 180.5 ± 3.7 cm; weight: 79.8 ± 4.4 kg), by dropping down from the top of a wooden box of 31 cm height, performed three one-leg landing tasks featured by different landing directions: frontward (FL), lateral (LL), and rotational (RL). The trunk and lower limbs' joints biomechanical responses were assessed through a laboratory setup consisting in a 6-camera motion capture system synchronized with force plates. A repeated measures one-way Analysis of Variance (ANOVA) was used to investigate the main effects of the landing directions on the lower limbs joints biomechanics during the main phases of interest of the landing tasks.

Results: The results indicated two major findings: (1) FL resulted in an advantageous active whole body configuration; (2) The lower limbs' joints adopted different and specific energy absorption strategies between the landing tasks.

Conclusions: This study indicates that the different landing tasks likely generate meaningful changes in whole body biomechanics and specific lower limbs joints kinetic and kinematic responses. Such outcomes might be used by clinicians towards a practical approach in selecting either assessment modalities or exercise interventions.

Keywords: motion capture, laboratory assessment, energy absorption, injury risk

INTRODUCTION

Landing maneuvers represent complex tasks concerning many areas of research, due to the variety of motor mechanisms and mechanical responses involved for their execution. The ability to land properly is required in many locomotor human activities, e.g., after sudden unexpected falls,¹ while stepping down stairs,² or in jumping-down situations.^{3,4} Research interest has also focused on specific applications among dynamic environments related to various intensive high-risk sports, whereas the inherent practical goal is to improve the quality of the landing performance in order to prevent injuries. In fact, landing-related injuries are common in sports,³ such as in volleyball,⁵ handball, soccer, basketball,⁶ and gymnastics.⁷ In these activities, single-leg landings, are often performed, and the forceful landing impact is primarily attenuated by the lower extremity joints' muscle actions. These actions serve to dissipate kinetic energy, converted from the potential energy of the athlete at a particular landing height.⁸ Much of the research on sport-related injuries due to landing events has focused on the pathomechanic aspects potentially related to injury occurrence.⁹ High magnitude of the ground reaction forces (GRFs) and abnormal external moments acting around the lower limbs' joints, have been associated with lower extremity injury.^{3,6,9} The majority of the biomechanical studies of landing attempted to quantify and compare the specific risk of injury between genders,¹⁰ between pre-injured and controlled populations, or between fatigue and non-fatigue conditions.¹¹ However only few studies have investigated landing tasks performed with various landing directions¹² which reproduce real-game situations commonly encountered in sport contexts.^{13,14} Wikstrom et al.,¹⁵ reported poor postural stability during diagonal and lateral landing tasks compared with forward direction. Sinsurin and colleagues¹² suggested that the risk of lower limb injury increase as the landing task direction changes from forward to lateral as a result of a stiff landing strategy adopted. Furthermore, Cowling and Steele,¹⁶ found that requiring an athlete to catch a ball during

flight altered hip and trunk sagittal plane kinematics during single leg landing. Therefore, a landing task that has multiplanar components, which better reflect the game scenarios, is related to high knee loading and increased ACL injury risk. It appears that due to the different mechanical demands of diagonal, lateral and rotational landing directions, whole body postures result in an extended and internally rotated lower limb, with an abducted hip, which likely increase the injury risk.¹⁴ Although these evidences, there is yet to be a study investigating the relationship between landing tasks performed through the three planes of motion toward different directions and the three-dimensional mechanical responses of lower extremities. An understanding of how full body motion affect lower limbs biomechanics is important, particularly in team sports that requires athletes to repeatedly perform such landing tasks during games and practice. Indeed, the results of this study could provide clinicians, rehabilitators, and sport professionals with useful indications about the use of optimal motion strategies for injury prevention and training purposes.

In reviewing the current literature and considering the above evidence-based framework, the aim of this study was to investigate differences in the biomechanical responses between various landing tasks performed toward different directions. It was hypothesized that different landing directions would result in changed GRFs and moments responses of lower extremity joints when attenuating the landing impacts. Secondly, we tested the hypothesis that variations in landing directions cause different and specific whole body postures for completing the landing tasks.

MATERIALS AND METHODS

Subjects

Based on the recommendations of Forrester,¹⁷ and assuming a large effect size (greater than 0.80), a sample size larger than 20 subjects performing three trials was needed to achieve a

power greater than 0.80 for determining differences among the experimental landing tasks. Experimental data were collected from a group of male subjects ($n = 30$), students of Sports Sciences (age: 26.4 ± 4.4 years; height: 180.5 ± 3.7 cm; weight: 79.8 ± 4.4 kg). Criteria for inclusion were: 1) healthy male subjects aged between 21 and 35 years; 2) no previous injuries or surgery in the back or in the upper or lower limb joints. The participants were given a soft copy of the volunteer information sheet duly approved by the University of Dundee Research Ethics Committee (Application Number: UREC 15112) according to the Declaration of Helsinki. On the appointed day for data collection, written informed consent was obtained from them after they received an oral explanation of the purpose, benefits, and potential risks of the study.

Experimental Approach

The experimental setup consisted of a 6-camera motion analysis system (Vicon® T-Series - T10S), synchronized with two Kistler force plates (Kistler Instruments AG, Winterthur, Switzerland) operating at a sampling frequency of 240 Hz and 960 Hz, respectively. Integrated Vicon Nexus software was used to synchronize the Vicon motion capture system and the Kistler force plates' data for the calculation of the biomechanical responses. The participants, wearing black spandex shorts and self-selected sport shoes, were instrumented with thirty-seven retroreflective markers of 12 mm in diameter at fixed locations and bony landmarks.¹⁸

Procedures

To ensure adequate familiarization, each participant was provided with a CD-ROM, including full instructions, and was asked to practice the experimental landing tasks. On the appointed day for data collection, before applying the retroreflective markers, some anthropometric and anatomical measurements were taken according to the Vicon®

manufacturer instructions. After the completion of the static calibration process and before the data collection, the participants performed a general warm-up, including 10 min of cycling at a constant intensity of 60 Watts (model 864, Monark, Inc, Stockholm, Sweden). The landing tasks (Figure 1) were performed by jumping down with one leg from the top of a wooden box of 31 cm height and placed 20 cm from the edge of the force plate.¹⁸ The participants stood with one leg on top of the box and then dropped off the box and landed onto the target area with the same leg. The landing tasks were designed as three types, and were arranged in a random order to avoid any order effect; a rest period of 1 min between the trials was preset to minimize the potential effects of fatigue. All data were sampled from the dominant limb (i.e., defined as the preferable leg reported to be used for jumping and landing with a single leg or during tasks of stepping down stairs). Participants performed three trials for each landing task.

Landing Tasks

The three landing tasks were defined as:

1. Frontward landing in the sagittal plane (FL) (Figure 1a); the participant was required to land following a frontward drop down while moving both arms, with elbows bent and swinging naturally, in the same direction of the landing.
2. Lateral landing in the frontal plane (LL) (Figure 1b); the participant was required to land following a sideward drop down while moving both arms with elbows bent naturally and in the same lateral direction of the landing.
3. Rotational landing in the horizontal plane (RL) (Figure 1c); the participant was required to land following a 90° rotational drop down while moving both arms with elbows bent naturally and in the same direction of the rotational landing.

In situations where the trials did not meet these requirements, the trial was discarded and not used for the data analysis, post-processing, or reporting.

***** Figure 1 about here *****

Data Processing and Analysis

Three-dimensional biomechanical motion data were processed with a custom-made MATLAB (R2015b) program developed for this study. Firstly, GRF data were filtered through a fourth-order, low-pass digital filter, with a cutoff frequency of 100 Hz, while markers' trajectories were filtered at a cutoff frequency of 12 Hz for kinematic calculations.¹⁸ Trunk segment and joint angular positions were obtained by using a Euler angles method in an "XYZ" rotation sequence (X for frontward-backward and flexion-extension, Y for side-flexion and adduction-abduction, Z internal-external rotation) based on a right-hand convention. Ankle angles were defined as the angles between the leg segment and foot segment. Knee angles were defined as the angles between the thigh and leg segments, while hip angles were defined as the angles between the thigh and pelvis segments. Trunk angles were calculated as the angles between the trunk segment and a vertical line in the laboratory coordinate system. Instantaneous joint angular velocities (ω) were then determined as the first derivative of angular position. All moments were reported as external joint moments derived from the GRFs created during contact with the force plates. In addition, sagittal hip-, knee-, and ankle-joint power curves (P) were calculated as ω and the net joint moment (M) ($P = \omega \cdot M$). The negative portions of the joint power curves were integrated to calculate negative mechanical joint work, representing the eccentric action required for energy absorption (EA) until stabilization. Total negative lower extremity work was calculated by summing up the negative joint works at the hip, knee, and ankle. Finally, the EA relative contribution of each

joint was determined.¹⁹ All the kinetic variables were normalized by the participants' body mass (N/kg, N·m/kg). For data analysis, each individual participant was represented by the mean of his trials.

Landing Phase Definitions

In order to analyze the measured data, the landing motion was investigated considering three key points and two phases:

- 1) Initial contact (IC): the time point corresponding to the first contact of the foot with the force plate, where the vertical GRF first exceeded a value of 10 N.²⁰
- 2) The peak of the GRF (GRF_{peak}): the time point corresponding to the peak value of the GRFs' responses.
- 3) Stabilization (S): the time point corresponding to a steady GRF value equal to the subject's body weight.

In addition, the time frame occurring between the IC and the GRF_{peak}, and between the GRF_{peak} and the S, were considered for determining the early phase and the terminal phase of the landing tasks, respectively. For these two phases, the EA contribution of each joint was calculated for the between-tasks analysis.^{21,22}

Statistics

The statistics software package SPSS (v22.0; IBM) was used to analyze the data, and statistical significance was set at $P \leq 0.05$. Descriptive statistics consisted of the means and standard deviations. The Intra-class Correlation Coefficient (ICC) was used to determine the reliability of the measures across the three trials performed. A repeated measures one-way Analysis of Variance (ANOVA) was used to investigate the main effects of the landing tasks on the lower-limb joints' biomechanics during the main phases of interest of the landing tasks. A post-hoc Bonferroni test was used when significant differences were detected, to

determine which of the dependent variables differed significantly between the experimental tasks. Effect size is presented as partial η^2 (eta squared).

RESULTS

All the variables were reasonably reliable, with ICC ranging from 0.811 and 0.842 for kinematic parameters, from 0.888 and 0.915 for kinetic variables, from 0.695 to 0.725 for joints' power measures, and from 0.895 to 0.922 for the GRF values.

Kinematic Responses

For joint position at IC, the FL task led to significantly greater ankle plantar-flexion ($F_{(2,29)} = 2397.12$, $p < 0.001$, partial $\eta^2 = 0.988$), knee flexion ($F_{(2,29)} = 1277.35$, $p < 0.001$, partial $\eta^2 = 0.978$), hip flexion angles ($F_{(2,29)} = 1368.35$, $p < 0.001$, partial $\eta^2 = 0.979$), and trunk frontward excursion ($F_{(2,29)} = 1480.25$, $p < 0.001$, partial $\eta^2 = 0.981$) in comparison to the other two tasks. The same landing task induced larger trunk frontward ROM excursion at GRF_{peak} ($F_{(2,29)} = 1543.25$, $p < 0.001$, partial $\eta^2 = 0.982$) and during the whole task ($F_{(2,29)} = 397.55$, $p < 0.001$, partial $\eta^2 = 0.932$), respectively (Figure 2).

Frontal plane kinematic responses of both the lower-limb joints and the trunk were influenced by the landing tasks performed. In summary, the FL task produced lower abduction angles in the knee ($F_{(2,29)} = 393.45$, $p < 0.001$, partial $\eta^2 = 0.931$) and lower adduction angle in the hip ($F_{(2,29)} = 294.10$, $p < 0.001$, partial $\eta^2 = 0.921$) at IC. At GRF_{peak} point, lower knee abduction ($F_{(2,29)} = 257.10$, $p < 0.001$, partial $\eta^2 = 0.899$) and trunk side flexion angles ($F_{(2,29)} = 74.55$, $p < 0.001$, partial $\eta^2 = 0.720$) were reported for the same task. Finally, the frontal plane ROMs of the knee ($F_{(2,29)} = 764.25$, $p < 0.001$, partial $\eta^2 = 0.963$) and hip joints ($F_{(2,29)} = 2345.05$, $p < 0.001$, partial $\eta^2 = 0.988$) and trunk segment ($F_{(2,29)} =$

1409.35, $p < 0.001$, partial $\eta^2 = 0.980$) were significantly lower for the FL with respect to the others (Figure 2).

In the transverse plane the main finding was that the FL induced lower peak trunk internal rotation angles at the IC and GRF_{peak} points, and ROM excursion ($F_{(2,29)} = 783.25$, $p < 0.001$, partial $\eta^2 = 0.964$, $F_{(2,29)} = 1460.15$, $p < 0.001$, partial $\eta^2 = 0.981$ and $F_{(2,29)} = 765.35$, $p < 0.001$, partial $\eta^2 = 0.963$, respectively) (Figure 2).

***** Figure 2 about here *****

Kinetic Responses

At IC, FL led to the highest ankle dorsi-flexion ($F_{(2,29)} = 211.25$, $p = 0.001$, partial $\eta^2 = 0.879$) and hip flexion moments ($F_{(2,29)} = 6629.15$, $p < 0.001$, partial $\eta^2 = 0.996$) than the other tasks. At the GRF_{peak} point, the same task produced higher ankle dorsi-flexion ($F_{(2,29)} = 449.55$, $p < 0.001$, partial $\eta^2 = 0.995$) and knee flexion ($F_{(2,29)} = 498.25$, $p < 0.001$, partial $\eta^2 = 0.945$) moments. Hip ($F_{(2,29)} = 16523.15$, $p < 0.001$, partial $\eta^2 = 0.998$) and knee ($F_{(2,29)} = 1203.15$, $p < 0.001$, partial $\eta^2 = 0.976$) flexion moments were notably greater for the FL in comparison to the other tasks at the S point (Table 1).

At both the IC and GRF_{peak} points, the FL task led to the lowest knee abduction ($F_{(2,29)} = 9858.30$, $p < 0.001$, partial $\eta^2 = 0.997$ and $F_{(2,29)} = 2385.25$, $p < 0.001$, partial $\eta^2 = 0.998$) and hip abduction ($F_{(2,29)} = 2874.20$, $p < 0.001$, partial $\eta^2 = 0.990$ and $F_{(2,29)} = 2186.45$, $p < 0.001$, partial $\eta^2 = 0.987$) moments in comparison with the other tasks as reported in Table 1.

At GRF_{peak}, FL led to the lowest values of knee and hip internal rotation moments ($F_{(2,29)} = 399$, $p < 0.001$, partial $\eta^2 = 0.932$ and $F_{(2,29)} = 937.45$, $p < 0.001$, partial $\eta^2 = 0.970$, respectively) (Table 1).

*** Table 1 about here ***

Energy Absorption

The ankle and the hip joint displayed the highest EA contributions ($F_{(2,29)} = 399$, $p < 0.001$, partial $\eta^2 = 0.932$ and $F_{(2,29)} = 937.45$, $p < 0.001$, partial $\eta^2 = 0.970$, respectively) during the early phase for the FL task as seen in Figure 3. As for the outcomes found during the terminal phase, the knee ($F_{(2,29)} = 28.273$, $p = 0.01$, partial $\eta^2 = 0.672$) and hip ($F_{(2,29)} = 532.27$, $p < 0.01$, partial $\eta^2 = 0.948$) joint EA contributions results were higher for FL when compared to the other two tasks (Figure 3).

*** Figure 3 about here ***

Ground Reaction Forces

The FL task led to the lowest values of GRFz at IC and as peak values ($F_{(2,29)} = 14843.17$, $p < 0.001$, partial $\eta^2 = 0.998$, $F_{(2,29)} = 536.25$, $p < 0.001$, partial $\eta^2 = 0.949$, respectively). However, the same task led to the highest values of GRFx both at IC and as peak values ($F_{(2,29)} = 2340.27$, $p < 0.001$, partial $\eta^2 = 0.988$, $F_{(2,29)} = 1748.15$, $p < 0.001$, partial $\eta^2 = 0.984$, respectively). Finally, in terms of GRFy, the FL led to the lowest values at IC and as peak values ($F_{(2,29)} = 2837.26$, $p < 0.001$, partial $\eta^2 = 0.990$, $F_{(2,29)} = 220.45$, $p < 0.001$, partial $\eta^2 = 0.883$, respectively) (Table 2).

*** Table 2 about here ***

DISCUSSION

The purpose of this study was to investigate the differences in three-dimensional biomechanics among three single-leg landing tasks performed toward different directions.

The results indicated two major findings: (1) The differences in the magnitude of biomechanical responses of the lower-limb joints could likely be influenced by the direction of the landing task performed; (2) The ankle and hip joints were the top contributors to EA during the early-phase when the single-leg landing was performed as FL task while during the terminal phase of the landing, the knee and hip joints acted as the key energy dissipaters in response to the FL task and in comparison with the other tasks.

The primary finding was that FL highlighted an advantageous whole body configuration exhibiting specific biomechanical responses. Previous evidence on the coupling mechanisms between the lower limbs` joints suggests that a more erect posture during closed-kinematic-chain tasks, as identified by more extended trunk, hip, and knee positions, may be a risk condition for anterior cruciate ligament (ACL) injury.²³ Accordingly, Devita and Skelly²⁴ found that the impact in an upright posture (stiff landing) was greater than that in a flexed posture (soft landing), as a consequence of the muscular system's inability to supply the required EA generated in response to the contact with the ground. Specifically, during landings that are aimed at absorbing impact (i.e., soft landings), the trunk and lower limb segments rotate in a greater sagittal ROM excursion than during stiff landings.²⁴ Considering that actively rotating the trunk forward during landing simultaneously increased the peak hip and knee flexion angle during the loading phase, it is reasonable that trunk forward rotation exerts an indirect influence on ACL loading. These results are in agreement with previous evidences, confirming that the combination of more flexed lower-limb joints and greater forward-downward inclination of the trunk segment at IC, associated with wider ROM of the same body segments in the sagittal plane, influence the GRFs and kinetic responses.²⁵ In fact, the FL led to the lowest values of vertical GRF at IC and in terms of peaks (Table 2). These lowest vertical GRF responses may be due to the larger involvement of the trunk, hip, knee, and ankle muscles, which are better suited to absorb the mechanical impact energy in

comparison to the same task performed as LL and RL. In fact, the extent of EA reflects the level of mechanical demands placed on the lower extremity and the corresponding mechanical responses of lower extremity musculature during landing.¹³ Therefore, increased EA would be beneficial, as the lower extremity joints muscles, acting in an eccentric mode, are able to provide adequate shock absorption without over-stressing the vulnerable soft tissue structures such as the ligaments.

An important finding arising from this study was that both the frontal and transverse planes' biomechanical responses were influenced by the landing task performed (Figure 3). From a mechanical perspective, EA in the frontal and transverse planes could be maximized similarly to that of the sagittal plane through specific planar joint motions. However, due to an abnormal lower-limb and trunk segments' configuration at landing, the GRF vector may move and consequently have a greater lever arm relative to the knee joint center, which in turn may lead to detrimental mechanical responses. This evidence relies on the notion of the "position of no return",²⁵ during which non-contact ACL injury is hypothesized to occur due to abnormal biomechanical responses involving the trunk and lower-limb joints.^{27,28} Greater knee abduction angles and abduction moment, and 20% higher GRF were identified as potential risk factors for ACL injury. Hewett et al.,²⁹ further confirmed that increased valgus motion and moments at the knee joint during landing impact are key predictors of a heightened potential for ACL injury. Moreover, hip frontal kinematics represent an important determinant of impact force at the knee during landing as well. Zhang et al.,³⁰ have previously reported a general increase in hip moments and powers in frontal plane biomechanics, which indicated that the hip abductor muscles can substantially enhance their EA capacity in the frontal plane during landing at increasing heights. In this study, it was noted a characteristic pattern in the frontal plane biomechanics, with: 1) **lowest horizontal medio-lateral peak GRF for FL and highest horizontal medio-lateral peak GRF for the LL and RL landing**

tasks; 2) higher peak adduction angles and frontal hip ROM for the LL and RL landing tasks in comparison with the FL; 3) higher values of hip external abduction moments both at IC and during the early phase of landing in response to the LL and RL tasks when compared with the FL; and 4) lower abduction angle values both at IC and as ROM during FL in comparison to the other tasks. Such results clearly demonstrate that the subjects adopted a hip-dominant strategy for EA in the frontal plane, and that the hip-joint muscles' contribution to the shock absorption was influenced by the specific landing task required. This suggests that different joints may have prominent EA capacities in specific planes; the ankle, knee, and hip joints might provide EA in the sagittal planes at different points of the landing tasks, while the hip joint contribution may be superior during proper closed-kinetic-chain task execution in the frontal plane.^{24,30,31}

The results of this study also showed specific patterns for the frontal trunk kinematics. LL and RL resulted in higher peak values of lateral trunk flexion (toward the direction of the landing leg) and greater trunk frontal ROM excursion. In addition, the concomitant presence of lateral trunk flexion, higher hip adduction, and knee abduction angles (Figure 2) highlights a possible coupling between the upper body and lower limbs in the frontal plane responses of closed-kinematic-chain tasks. FL was featured by lower values of peak and ROM trunk lateral flexion, together with lower hip adduction and knee abduction moments both at IC and during the whole execution of the landing task. On the other hand, LL and RL led to greater frontal plane kinetic responses, highlighted by higher hip adduction and knee abduction moments in comparison with the FL. These findings demonstrate, on the basis of experimental data, the entire coupling between the trunk, hip, and knee during landing maneuvers performed along specific directions.

In accordance with the above mechanical outcomes, similar trends of the lower-limb joints' EA contributions were observed when comparing the three landing tasks. Ankle and

hip joints were the top contributors to EA at the early-phase when the single-leg landing was performed as FL, while during the terminal phase, the knee and hip joints acted as the key energy dissipaters (Figure 3). These findings suggest that the ankle plantar flexors, knee extensors, and hip extensors were working *harder* to provide adequate EA in response to the FL task. However, different strategies of energy absorption between the three lower-limb joints were observed during the execution of the landing tasks. Specifically, the EA data suggest that the ankle and the hip joints provided the higher contribution at the early phase, while the ankle joint was less capable of EA compared with the hip and knee joints, during the terminal-phase of the landing task (Figure 3). These results confirm the role of the ankle joint in landing mechanics and force attenuation at IC, to reduce potential injury-causing peak vertical GRF and loading rates. Kovacs et al.,³² found that vertical GRFs were significantly higher in heel-toe landings than in forefoot landings. The ankle plantar-flexion dissipated the majority of the energy in the forefoot landing, but were second to the knee and hip extensors during the whole task as confirmed by the results of this investigation. It has been suggested that the direction of power transfer during landing was from the distal to proximal segments;³²⁻³⁴ it is therefore logical to assume that the energy generated before landing contact can be transported from the distal end to the more proximal and massive muscle group for further dissipation during the later phases of the landing tasks.

An interesting direction for future research could be to set up assessment tests with ecological validity by adding sport-specific elements that more resemble real game situations (e.g., an overhead target and lateral force disturbance before landing).^{35,36} Indeed, it would be interesting to assess the effects of landing directions on the lower extremity biomechanics in different athletic populations. Future studies should also include electromyography to examine the individual contributions of the torso muscles in trained subjects and to

understand the role of the torso musculature in reducing knee loading and in helping to prevent injury.

CONCLUSIONS

In conclusion, different landing directions likely generate meaningful changes in whole body biomechanics and in specific kinetic and kinematic responses. In comparison to the LL and RL landing tasks, during the FL the subjects assumed an advantageous whole-body configuration leading to a soft landing and resulting in lower GRFs and lower extremities joints moments responses. In addition, the results suggest that different joints may have prominent energy dissipation capacities in specific planes. The ankle, knee, and hip joints might may provide energy dissipation in the sagittal planes at different points of the landing tasks, while the hip joint contribution may be superior during the task execution in the frontal plane.

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NOTES

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TITLES OF FIGURES AND TABLES

Table 1. Kinetic responses across the three landing phases

Table 2. Ground reaction forces responses

Figure 1. Experimental landing tasks: FL (a), LL (b), RL (c).

Figure 2. Three-dimensional kinematical responses at the three points of the landing tasks: IC (upper graphs), GRF_{peak} (midline graphs), S (lower graphs). The values are shown as mean and standard deviation. *Note.* “*” indicates significant differences from FL; Statistical significance level was set at $p \leq 0.05$.

Figure 3. Lower-limbs joint EA contributions at the two phases of the landing tasks: early-phase (left side graph), terminal-phase (top side graph). *Note.* “*” indicates significant differences for the ankle joint; “#” indicates significant differences for the knee joint; “\$” indicates significant differences for the hip joint; Statistical significance level was set at $p \leq 0.05$.

a



b



c



